The Dynamics of Gait Transitions: Effects of Grade and Load

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ABSTRACT. Diedrich and Warren (1995a) proposed that gait transitions behave like bifurcations between attractors, with the relative phase of the leg segments as an order parameter and stride frequency and stride length as control parameters. In the present experiments, the authors tested the prediction that manipulation of the attractor layout, either through the addition of load to the ankles or through an increase in the grade of the treadmill, induces corresponding changes in the walk-run transition. As predicted, the load manipulation shifted the most stable walk and the transition to lower stride frequencies. In contrast, the grade manipulation shifted the most stable walk and the transition to shorter stride lengths. Other features of the dynamic theory were also replicated, including enhanced fluctuations of phase and systematic changes in stride length and frequency at the transition. Overall, in these experiments a shift of the attractors in control parameter space yielded a corresponding shift of the transition.

Key words: dynamics, gait transitions, human locomotion

Why do humans and other animals switch from one stable pattern of coordination to another as conditions change? Gait transitions provide a universal case in point, and in the present studies we focused on the shift between walking and running in humans. According to a standard view in psychology, the walk–run transition is caused by the switching of central motor programs that prescribe the movement pattern for each gait (e.g., Shapiro, Zernicke, Gregor, & Diestel, 1981). That explanation, however, tells us little about why gait transitions occur and what the expected behavior is at gait shifts. In contrast, biomechanics researchers seek to account for the particular transition speeds that are observed. Although there is little agreement as to the specific processes responsible for the transition, numerous explanations have been proposed, including that the transition occurs at the maximum possible walking speed (e.g., Alexander, 1984); that the transition minimizes the chance of injury (Biewener & Taylor, 1986; Farley & Taylor, 1991; Hreljac, 1995); metabolic energy expenditure (e.g., Mercier et al., 1994), or mechanical work (Alexander, 1980, 1992); or that it optimizes a comfort factor related to muscle efficiency (Minetti, Ardigo, & Saibene, 1994a; for a review, see Diedrich & Warren, in press).

An alternative approach argues that preferred motor patterns and the transitions between them arise from the self-organizing dynamics of the action system (Kelso, 1995; Kugler & Turvey, 1987) and, specifically, that the shift between gaits behaves as a bifurcation between two attractors (Diedrich & Warren, 1995a, in press). In the present study, we pursued such a dynamic account of the walk–run transition. Because the theory proposes that gait transitions result from competition between attractors for each gait, it follows that shifting the loci of the attractors should induce a similar shift in the transition point. In two experiments, we manipulated the walking and running attractors by either loading the ankles or increasing the grade of the treadmill and looked for corresponding changes in the walk–run transition.

A Dynamic Approach to the Walk–Run Transition

A dynamic approach to locomotion views preferred gaits and gait transitions as arising from the dynamics of the locomotor system (Collins & Richmond, 1994; Collins & Stewart, 1993; Schöner, Jiang, & Kelso, 1990; Taga, 1995a, 1995b; Taga, Yamaguchi, & Shimizu, 1991; Thelen & Ulrich, 1991; Turvey, Holt, Obusek, Salo, & Kugler, 1996;
Wagenaar & van Emmerik, 1994). Following work by Kelso and his colleagues on phase transitions in bimanual coordination (Haken, Kelso, & Bunz, 1985; Kelso, Scholz, & Schöner, 1986; Schöner, Haken, & Kelso, 1986), we proposed that the walk–run transition behaves as a bifurcation between attractors in relative phase for each gait (Diedrich & Warren, 1995a). In particular, we suggested that the order parameter characterizing the organization of human gait is the relative phase of the segments within a leg, whereas the nonspecific control parameters that scale the system through a transition are stride length and stride frequency. Relative phase between the legs does not reflect changes between bipedal gaits; therefore, the relative phase of the segments within a limb is a good candidate variable for the order parameter because it captures the spatiotemporal patterning of limb movement (see Kelso, Buchanan, & Wallace, 1991). Because speed, v, is the product of stride frequency, f, and stride length, s (i.e., v = fs), and stride length and frequency are systematically coupled during normal locomotion, speed often behaves as a single control parameter (Diedrich & Warren, 1995a).

The hypothesized dynamics of the walk–run transition, in which we assumed a potential function V(x) of the form of Equation 1,

\[ V(x) = kx - x^2/2 + x^4/4, \]  

is illustrated in Figure 1, where k is the control parameter and x is the order parameter in arbitrary units (shown here, k ranges from -1 to 1, x ranges from -3 to 3; see Tuller, Case, Ding, & Kelso, 1994). For simplicity, we assumed that stride frequency and stride length are coupled as a single control parameter (speed). Referring to the solid curve, as speed increases (or decreases) one mode of behavior becomes unstable and the system bifurcates to the more stable mode, yielding a gait transition. It follows that a gait transition should have the following properties (Kelso & Schöner, 1988): (a) There should be a qualitative and sudden reorganization of the system at the transition (a bifurcation), as reflected by a change in the relative phase of the segments within a leg. (b) There should be a tendency for the system to remain in the current basin of attraction as the control parameter moves the system through the transition region. Therefore, the system should often exhibit hysteresis, which is a tendency to make the walk-to-run (W–R) transition at a higher speed than the run-to-walk (R–W) transition (see Alexander, 1989). (c) The locomotor system should show a loss of stability in the transition region, as indexed by critical fluctuations and critical slowing down. Critical fluctuations occur as the current basin of attraction broadens near the transition: Given a constant level of noise, the system will occupy more states, thus yielding enhanced fluctuations. For human gait, fluctuations are indexed by an increase in the standard deviation of the relative phase of the leg segments. Critical slowing down in the transition region occurs for a similar reason: As the basin of attraction broadens near the transition, its gradients become shallow and it takes more time for the system to recover from a perturbation.

In Diedrich and Warren (1995a), data were reported for the walk–run transition in humans that were consistent with a dynamic model. First, on trials in which speed was continuously scaled and the participants were allowed to make the transition freely, we observed that the transition was characterized by a qualitative reorganization of the relative phase of the leg segments, which occurred within one stride cycle (Figure 2), and that the W–R transition tended to occur at a higher speed than the R–W transition (see also Beuter & Lalonde, 1989; Hreljac, 1993, Thorstensson & Robertsson, 1987). Second, on trials in which participants were held in specified gaits at various constant speeds, we observed a qualitative difference in relative phase between gaits and an increase in fluctuations of relative phase near the typical transition region. Although we did not directly measure critical fluctuations by testing stability in free transition trials, the steady-state method allows one to reliably measure phase fluctuations in a given gait at speeds below, at, and above the transition (for more on steady-state methodology, see Schmidt, Shaw, & Turvey, 1993; Schmidt & Turvey, 1995). Those data demonstrated that the control parameter (speed) weakens the dynamics as it is scaled, just as one would expect if relative phase provides an index of different attractor states for walking and running. Furthermore, because there was a sudden, nonlinear change in relative phase at the transition, from values that define a walk to values that define a run, we obtained strong evidence that the transition behaves as a bifurcation between two attractors in relative phase. We have yet to investigate critical slowing down, which requires a mechanical perturbation.

**Dynamics and Energetics**

Diedrich and Warren (1995a) speculated that the underlying dynamics of gait are also manifested in the overall energetics, because total metabolic cost reflects the consequences of driving the system away from its attractor states (along with other metabolic factors). Consistent with that notion, it has been found that gait transitions act to reduce both the internal mechanical work done in accelerating the limb segments and the total energy expenditure in overground locomotion (Minetti et al., 1994a). If that is the case, we may, conversely, use energy expenditure as a rough guide to the underlying dynamics.

Existing metabolic data, in the form of a contour plot of energy expenditure per unit distance (cal/kg/m) as a function of stride length and stride frequency, are summarized in Figure 3 (see Diedrich & Warren, 1995a, for details on the data used to construct this plot, following Molen, Rozendal, & Boon, 1972a, 1972b). In walking, there is a particular combination of stride length and frequency at which energy expenditure per unit distance is minimized (smallest circle in Figure 3), corresponding to an energetically optimal speed of about 1.3 m/s. Any change in stride frequency and length away from those values results in increases in ener-
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**FIGURE 1.** Hypothetical potential function $V(x)$ for the walk–run transition, where speed ($v$) increases over panels. Normal condition (solid curve): As speed increases, the system moves from a stable walking attractor ($v = 2$) into an unstable transition region ($v = 5$) and suddenly jumps to a stable running attractor ($v = 6$). Transformed condition (dashed curve): Walking attractor has moved to a lower speed ($v = 1$), accompanied by the unstable region ($v = 4$) and the jump ($v = 5$).

In contrast, for running there is no single speed that minimizes metabolic cost: Energy expenditure per unit distance is constant across a wide range of speeds. However, a departure from the preferred stride length–frequency combination at any given speed (middle set of parallel lines in Figure 3) results in increased energy expenditure (outer parallel lines in Figure 3). Those data suggest that there is a “valley” of attraction for running (parallel lines in Figure 3). Consistent with that claim, we have found that fluctuations of rel-

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Energy expenditure (larger circles in Figure 3). Such data suggest that there is a basin of attraction for walking, defined in control parameter space (concentric circles in Figure 3). Consistent with that claim, we found a similar U-shaped function for fluctuations of relative phase as a function of speed (Diedrich & Warren, 1995a), and U-shaped fluctuation functions have also been observed at a single speed as stride frequency and stride length were varied (Holt, Jeng, Ratcliffe, & Hamill, 1995).
Gait transitions can be accounted for in terms of competition between the two attractors defined in control parameter space. An equal-energy separatrix divides the regions of attraction (bold curve in Figure 3). Because the cost of a preferred run is approximately 1.0 cal/kg/m at any speed, the separatrix corresponds to those stride length–frequency combinations for which the energy cost of walking is also 1.0 cal/kg/m. Diedrich and Warren (1995a) found that the walk–run transition occurred near the separatrix at approximately 2.1 m/s, even when stride length and frequency were dissociated. Furthermore, the directions of change in stride length and stride frequency observed at the transition are predicted by a jump from one attractor to another (bold dashed line in Figure 3).

Manipulating the Attractors: The Effects of Grade and Load

A consequence of the dynamic theory is that manipulation of the attractor layout in control parameter space should be accompanied by corresponding changes in the equal-energy separatrix and the walk–run transition. Specifically, the theory predicts that a shift of the attractors along the stride length or stride frequency axes, or both, shown in Figure 3, will result in a corresponding shift of the transition. For instance, under normal conditions (solid curve in Figure 1) walking is most stable at Speed 2 in the area where the gradient of the potential well is steepest, and becomes unstable at Speed 5 in the area where the potential function broadens, allowing for the W–R transition. In the transformed condition (dashed curve), the walking attractor shifts down to Speed 1, reducing the transition region to Speed 4. Thus, a shift in the attractors is accompanied by a shift in the transition. Such alterations in the attractor layout may be induced by changes in the same variables that are known to affect the energetics of locomotion, for example, by adding an external load or varying the grade, compliance, or frictional characteristics of the support surface. Here, we examined the effects of loading the ankles and increasing the grade of the treadmill.

First, consider the effects of loading the legs. It is known that energy expenditure increases with added weight on the ankles or feet when walking (Inman, Ralston, & Todd, 1981; Soule & Goldman, 1969) and running (Claremont & Hall, 1988; Martin, 1985; Myers & Steudel, 1985), and that the preferred combinations of stride length and frequency at a given speed change. In addition, loading the ankle increases...
the moment of inertia of the swing leg, resulting in a decrease in the natural frequency of gait (Turvey, Schmidt, Rosenberg, & Kugler, 1988) and, therefore, a decrease in the preferred walking frequency (Holt, Hamill, & Andres, 1990). Preferred stride length tends to increase during walking (Inman et al., 1981) or running (Martin, 1985) with loaded legs, although a decrease in stride length has also been reported for running (Claremont & Hall, 1988). Taken together, those data indicate that the attractors shift to different values of the control parameters (a lower stride frequency and possibly a higher stride length) when the legs are loaded. The dynamic theory predicts that the transition should exhibit a similar shift in stride length–frequency space.

As a second example, consider the consequences of locomoting uphill. In general, energy expenditure per unit distance increases as grade increases during walking (Bobbert, 1960; Margaria, 1938, 1976) and running (Heinert, Serfass, & Stull, 1988; Margaria, 1938, 1976), accompanied by changes in the preferred stride length and frequency combinations. For walking, the speed requiring minimum energy expenditure per unit distance decreases from 1.30 m/s on a 0% grade to 1.26 m/s on a 10% grade. Most evidence indicates that stride length decreases when one locomotes uphill at comparable walking (Cotes & Meade, 1960) and running (Heinert et al., 1988; Minetti, Ardigo, & Saibene, 1994b; Nelson & Osterhoudt, 1971) speeds, although a slight increase in stride length has also been reported for walking (Minetti, Ardigo, & Saibene, 1993). Taken together, those data suggest that the attractors shift to lower speeds during uphill locomotion, probably because of a drop in stride length. The dynamic theory thus predicts that the transition will also shift to a lower speed, because of changes in stride length. Consistent with that claim, as grade increases from 0% to 10%, the equal-energy separatrix drops from 2.32 m/s to 1.90 m/s. Indeed, while we were performing the present experiments, it was reported that the walk–run transition, likewise, shifts to a lower speed when humans travel uphill (Hreljac, 1995; Minetti et al., 1994a), although the component changes in stride frequency and stride length were not reported.

In the experiments reported here, we manipulated load and grade in order to measure shifts in both the walking attractor (estimated by the minimum of fluctuations of relative phase) and the walk–run transition to different values of the control parameters. We did not measure the running attractor because running for long periods under load and grade conditions is difficult. Rather, on the basis of the above data, we assumed that the walking and running attractors move in a like manner, although it remains possible that other variables may differentially affect the two gaits.

**EXPERIMENT 1**

**The Effects of Load**

In Experiment 1, the effects of loading the ankles were examined in two conditions: unloaded (0 kg) and loaded (2.27 kg on each leg). In *transition trials*, participants walked or ran on a treadmill while its speed was continuously varied, allowing us to analyze the transition behavior. In *plateau trials*, we analyzed each participant’s steady-state behavior at a variety of constant speeds while he or she remained in a particular gait (speed changed stepwise, in an ascending or descending order), which allowed us to identify the most stable speed, stride frequency, and stride length. We hypothesized that loading the ankles would lead to a drop in the most stable stride frequency and, consequently, a decrease in the transition stride frequency as well.

**Method**

**Participants**

Eight people (4 men, 4 women) were paid to participate in the experiment. Individual participants’ characteristics are presented in Table 1.

**Apparatus**

Each participant’s leg position was sampled at 100 Hz by an ELITE motion analysis system (Bioengineering, Technology, and Systems, Milan, Italy), which tracked the movement of five passive infrared reflecting markers on the left leg. One marker was attached to the thigh, in line with the greater trochanter (hip) and the lateral femoral condyle (knee). Placement of the marker in that location allowed for measurement of the hip angle while avoiding occlusion by the swinging arm. A second marker was placed on the lateral femoral condyle, and a third marker was placed on the lower leg, in alignment with the lateral femoral condyle and the lateral malleolus (ankle), in a way that avoided the ankle weight but permitted measurement of the joint angles. Markers were also placed on the lateral aspect of the calcaneus (heel) and the lateral side of the fifth metatarsal (toe); for the definitions of the joint angles, which were sagittal plane projections of the angles between the thigh, lower leg, and foot segments, see Figure 4). The hip angle was measured with respect to the vertical, so it did not capture changes in the inclination of the trunk. Movements of the markers were recorded by two cameras placed on the participant’s left side; the three-dimensional (3D) reconstruction of marker position had an accuracy of 1.0 mm in the sagittal plane and 2.0 mm in the lateral plane.

We used foot switches (Tapeswitch Corporation, Model 121-BP) to measure the touchdown and liftoff of each foot. Two switches were used for each foot, one under the heel in the anterior-posterior direction and one under the metatarsals in the medial-lateral direction. The participants walked and ran on a Quinton Q55 motorized treadmill. The foot switches and an analog signal of the treadmill speed were sampled by the ELITE system at 100 Hz (see Diedrich & Warren, 1995a, for details).

In the loaded condition, participants wore 2.27-kg lead ankle weights on both legs. The weights were sewn into a
Trials were blocked by load condition, and condition order collected in the was counterbalanced across speed plateaus in both the unloaded and loaded trials. In each condition, there were four blocks of trials each, for a total of 40 trials per condition. Because of a software limit on data acquisition, 30 trials in each load condition were collected in the first session, whereas the remaining 10 trials in each condition were completed in the second session. Trials were blocked by load condition, and condition order was counterbalanced across participants, as was the direction of progression (ascending vs. descending). Each trial was 30 s in duration at one constant speed; data collection began approximately 10 s after adjustment of the treadmill speed, and each session lasted 1.5 to 2 hr.

During the second session, participants again warmed up by walking for 5–10 min in the relevant load condition and completed the remaining plateau trials. They then performed transition trials in both load conditions, preceded by the following instructions: “For these trials we will be changing the speed of the treadmill while you are on it. Please walk or run as feels comfortable. That is, make the transition when it seems natural to do so.” Transition trials consisted of a continuous increase (W–R) or decrease (R–W) of the treadmill speed through the range of 1.17 m/s to 3.0 m/s. Participants practiced 5 trials in the W–R direction and 5 trials in the R–W direction. Data were then collected for 10 trials in each direction, resulting in a total of 20 trials per load condition. Trials were again blocked by load condition and the order of transition directions (W–R vs. R–W) was counterbalanced across participants. Each trial was 30 s in duration.

It is important to note that the procedure followed in the plateau trials presumably conformed to the delay convention, \( t_{\text{relaxation}} < t_{\text{control}} < t_{\text{passage}} \) (Kelso, Ding, & Schöner, 1992; Schmidt, Carello, & Turvey, 1990). In general, the time required to recover from a perturbation, \( t_{\text{relaxation}} \), is likely to be very small and to take at most a few stride cycles (Diedrich, 1995; Kelso, Holt, Rubin, & Kugler, 1981; Shik & Orlovskii, 1965). That recovery time is less than the time scale of change in the control parameter, \( t_{\text{control}} \) (40 s in the present experiments). Finally, \( t_{\text{control}} \) is less than the time scale associated with the first passage from one gait to anoth-

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**TABLE 1**

Participants' Characteristics

<table>
<thead>
<tr>
<th>Part. no.</th>
<th>Exp.</th>
<th>Gender</th>
<th>Age</th>
<th>Wt. (kg)</th>
<th>Ht. (m)</th>
<th>Leg (m)</th>
<th>Segment (m)</th>
<th>Exercise (km/week)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>L</td>
<td>M</td>
<td>22</td>
<td>64</td>
<td>1.83</td>
<td>0.98</td>
<td>0.44 0.45 0.14</td>
<td>56</td>
</tr>
<tr>
<td>2</td>
<td>L, G</td>
<td>F</td>
<td>21</td>
<td>63</td>
<td>1.65</td>
<td>0.89</td>
<td>0.39 0.41 0.14</td>
<td>24</td>
</tr>
<tr>
<td>3</td>
<td>L, G</td>
<td>F</td>
<td>21</td>
<td>53</td>
<td>1.60</td>
<td>0.85</td>
<td>0.38 0.38 0.12</td>
<td>32</td>
</tr>
<tr>
<td>4</td>
<td>L, G</td>
<td>M</td>
<td>26</td>
<td>67</td>
<td>1.73</td>
<td>0.90</td>
<td>0.40 0.40 0.14</td>
<td>60</td>
</tr>
<tr>
<td>5</td>
<td>L</td>
<td>M</td>
<td>20</td>
<td>64</td>
<td>1.70</td>
<td>0.93</td>
<td>0.40 0.42 0.12</td>
<td>32</td>
</tr>
<tr>
<td>6</td>
<td>L</td>
<td>F</td>
<td>21</td>
<td>56</td>
<td>1.68</td>
<td>0.93</td>
<td>0.41 0.42 0.12</td>
<td>32</td>
</tr>
<tr>
<td>7</td>
<td>L</td>
<td>F</td>
<td>31</td>
<td>60</td>
<td>1.70</td>
<td>0.93</td>
<td>0.39 0.42 0.14</td>
<td>0</td>
</tr>
<tr>
<td>8</td>
<td>L</td>
<td>M</td>
<td>20</td>
<td>66</td>
<td>1.73</td>
<td>0.88</td>
<td>0.37 0.39 0.14</td>
<td>5</td>
</tr>
<tr>
<td>9</td>
<td>G</td>
<td>M</td>
<td>21</td>
<td>66</td>
<td>1.73</td>
<td>0.91</td>
<td>0.40 0.41 0.15</td>
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<tr>
<td>10</td>
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<td>F</td>
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<td>48</td>
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<td>0.37 0.38 0.10</td>
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<td>11</td>
<td>G</td>
<td>F</td>
<td>21</td>
<td>66</td>
<td>1.73</td>
<td>0.93</td>
<td>0.42 0.41 0.14</td>
<td>20</td>
</tr>
</tbody>
</table>

Note. In the column labeled Exp., L stands for the load experiment and G stands for the grade experiment. The participants had their shoes on when the leg length measurements were taken. Segment 1 is from the greater trochanter (hip) to the lateral femoral condyle (knee), Segment 2 is from the knee to the lateral malleolus (ankle), and Segment 3 is from the ankle to the lateral side of the fifth metatarsal (toe). In the last column, the average weekly running routine for each participant is shown. Participant 7 used a Nordic Track 3 days a week, Participant 9 played basketball 3 days a week, and Participant 10 danced 5 days a week.

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For instance, a participant with a leg length of 0.60 m walked at speeds ranging from 0.59 m/s to 2.30 m/s. In each load condition, there were four blocks of 10 trials each, for a total of 40 trials per condition. Because of a software limit on data acquisition, 30 trials in each load condition were collected in the first session, whereas the remaining 10 trials in each condition were completed in the second session. Trials were blocked by load condition, and condition order was counterbalanced across participants, as was the direction of progression (ascending vs. descending). Each trial was 30 s in duration at one constant speed; data collection began approximately 10 s after adjustment of the treadmill speed, and each session lasted 1.5 to 2 hr.

During the second session, participants again warmed up by walking for 5–10 min in the relevant load condition and completed the remaining plateau trials. They then performed transition trials in both load conditions, preceded by the following instructions: “For these trials we will be changing the speed of the treadmill while you are on it. Please walk or run as feels comfortable. That is, make the transition when it seems natural to do so.” Transition trials consisted of a continuous increase (W–R) or decrease (R–W) of the treadmill speed through the range of 1.17 m/s to 3.0 m/s. Participants practiced 5 trials in the W–R direction and 5 trials in the R–W direction. Data were then collected for 10 trials in each direction, resulting in a total of 20 trials per load condition. Trials were again blocked by load condition and the order of transition directions (W–R vs. R–W) was counterbalanced across participants. Each trial was 30 s in duration.

It is important to note that the procedure followed in the plateau trials presumably conformed to the delay convention, \( t_{\text{relaxation}} < t_{\text{control}} < t_{\text{passage}} \) (Kelso, Ding, & Schöner, 1992; Schmidt, Carello, & Turvey, 1990). In general, the time required to recover from a perturbation, \( t_{\text{relaxation}} \), is likely to be very small and to take at most a few stride cycles (Diedrich, 1995; Kelso, Holt, Rubin, & Kugler, 1981; Shik & Orlovskii, 1965). That recovery time is less than the time scale of change in the control parameter, \( t_{\text{control}} \) (40 s in the present experiments). Finally, \( t_{\text{control}} \) is less than the time scale associated with the first passage from one gait to anoth-

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er because of random fluctuations, \( \tau_{\text{passage}} \), for we observed no spontaneous gait transitions in any session, outside of the expected transition region.\(^5\)

**Data Analysis**

**Plateau trials.** We analyzed the plateau trials in order to measure stride frequency, stride length, a point estimate of the relative phase between the hip and ankle at push-off, and fluctuations of relative phase. At all speeds, means and within-trial standard deviations were determined from the first 16 complete stride cycles. Stride frequency was measured as the inverse of the period between successive activations of the left heel foot switch. Speed was determined from the treadmill output, averaged over a window of five frames, whereas stride length was computed by dividing the speed by the stride frequency (\( s = v/f \)).

All kinematic data were filtered with a fourth-order, Butterworth low-pass filter with an 11.5 Hz cut-off. Joint angle time-series, illustrated in Figure 5, were computed from those data. We used a peak-picking algorithm to locate peak extension of the hip and peak plantar flexion of the ankle near push-off. The algorithm located a joint angle extremum, searched forward and backward in time to find a point 1° less than the extremum (the noise criterion), and then defined the peak as the time average of those two side points. The peaks were used in a point estimate of relative phase, with peak plantarflexion of the ankle (target) defined as a proportion of the time between successive peak hip extensions (reference cycle), with a complete reference cycle being equal to 360° (Figure 5).\(^6\) Preliminary analysis in Diedrich and Warren (1995a) indicated that the phase relationships at both peak flexion and peak extension during stance were similar, thus validating the use of a single point estimate at peak extension. Because of the circular nature of the relative phase measure (360°), all means and standard deviations of phase were calculated by using the methods of circular statistics (Batschelet, 1981; Burgess-Limerick, Abernethy, & Neal, 1991; see also Diedrich & Warren, 1995a). Only the ankle–hip relationship was investigated, because previous results had indicated that the ankle–knee relationship behaves similarly with respect to fluctuations of relative phase in the speed range investigated in this study (Diedrich & Warren, 1995a).

**Transition trials.** We analyzed transition trials to obtain the stride frequency, stride length, and speed at the transition, and measured those variables in the same manner as we measured the plateau trials. The W–R transition was identified as the first stride containing a flight phase, whereas the R–W transition was defined as the last stride containing a flight phase. Transition values were taken from the end of the last complete stride cycle before the one containing the transition and corresponded to the pretransition values computed in Diedrich and Warren (1995a).\(^7\)

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**FIGURE 4.** Marker positions and joint angles for the left leg: 1 = thigh, 2 = knee, 3 = lower leg, 4 = ankle, 5 = toe, \( \alpha \) = hip angle (to vertical), \( \beta \) = knee angle, and \( \gamma \) = ankle angle.

**FIGURE 5.** Sample time-series of angular displacement of the hip and the ankle. The ankle (B, target) and hip peaks (As, reference cycle) used in the point estimate of relative phase are also shown.
Data from 19 of the 960 trials collected were missing because of equipment failure or marker occlusion. In the event of a missing trial, statistics were calculated from the remaining trials in that condition for that participant.

Results and Discussion

Plateau Trials

Preferred Stride Length and Frequency Combinations

The preferred combination of stride length and stride frequency at each prescribed walking speed is presented in Figure 6 (solid curve = 0 kg, dashed curve = 2.27 kg). Two-way repeated-measures analyses of variance (ANOVAs; Load x Speed) demonstrated that as speed increased, so did both mean stride frequency, $F(9, 63) = 652.98, p < .001$, and mean stride length, $F(9, 63) = 661.51, p < .001$. With the addition of a load, mean stride frequency decreased, $F(1, 7) = 14.25, p < .01$, whereas mean stride length increased, $F(1, 7) = 15.43, p < .01$. There was no Speed x Load interaction for either stride frequency, $F(9, 63) = 0.67, ns$, or stride length, $F(9, 63) = 1.05, ns$. Overall, those data suggest that ankle loading induced a shift of the walking attractor to different values of the control parameters, consistent with earlier findings. Because those measurements were made during a forced walk, however, they did not reveal the stride frequency-stride length combination used at the preferred walking speed or the precise direction of the shift in the attractor.

Relative Phase

As shown in Figure 7, the ankle and the hip tended to be somewhat out of phase at push-off during walking, such that the hip reached peak extension before the ankle reached peak plantarflexion. Watson–Williams $F$ tests for circular variables (Batschelet, 1981) indicated that mean relative phase decreased significantly as speed increased, $F(9, 70) = 3.35, p < .01$, thus replicating the drift in phase observed by Diedrich and Warren (1995a). Overall, there was no main effect of load, $F(1, 14) = 0.45, ns$, but the interaction was significant, such that ankle loading had the greatest effect at the lower speeds, $F(19, 140) = 3.81, p < .01$. Thus, the load manipulation reduced the mean relative phase at low speeds, indicating that the preferred phase of the walking attractor was affected.

Fluctuations of Relative Phase

The stability of gait was estimated from fluctuations in the phase relationship between the ankle and the hip (presented in Figure 8a). A Load x Speed ANOVA revealed that the standard deviation of relative phase varied significantly as a function of speed, $F(9, 63) = 12.56, p < .001$, and single-degree-of-freedom polynomial contrasts showed a significant quadratic component, $F(1, 7) = 47.95, p < .001$. That finding confirmed the U-shaped fluctuation function we had previously obtained with coarser sampling (Diedrich & Warren, 1995a), and its minimum indicated the
speed with the most stable gait. On the other hand, there was no effect of load, $F(1, 7) = 3.95, ns$, or Load x Speed interaction, $F(9, 63) = 0.63, ns$, apparently inconsistent with a shift in the minimum of the function. But the ANOVA is relatively insensitive, and cannot be applied to the stride length and frequency dimensions because they were not manipulated as factors. Therefore, we turned to a least squares curve-fitting procedure.

We analyzed fluctuations of phase separately as functions of speed, stride frequency, and stride length. First, for each load condition, we fit the data for mean standard deviation of phase as a function of speed with an equation of the form

$$y = \frac{A}{\sqrt{x(1-x/B)^2}} + C,$$

where $y$ is the standard deviation of phase, $x$ is Froude number, and $A$, $B$, and $C$ are constants. That equation was taken from the analysis by Inman et al. (1991) of energy expenditure during walking and consistently provided the best fit to our fluctuation functions when compared with a simple quadratic and a hyperbolic of the form $y = A/x + Bx + C$. It is important to note that the shifts reported in the following section did not depend on the specific equation chosen, for all of the equations tested showed shifts in the same directions. Mean standard deviation of phase was also plotted as a function of stride frequency and stride length, and all fits were done with Equation 3. Regression constants, $r^2$ values, and minima appear in Table 2.

The fits for fluctuations in phase as a function of speed appear in Figure 8a. Differentiating the regression equations indicated that the minimum of the function occurred at a Froude number of 0.31 (1.66 m/s) in the unloaded condition and 0.30 (1.64 m/s) in the loaded condition. Expressed as a percentage of the total range of variation in walking speed in all plateau trials, that corresponded to a 1.8% decrease in the most stable speed, indicating that the walking attractor may have shifted in speed slightly. More important, the minimum of the fluctuation function shifted along the stride frequency axis from 1.03 Hz in the unloaded condition to 0.98 Hz in the loaded condition (Figure 8b), a shift of 7.6% of the observed frequency range. In contrast, there was no such shift along the stride length axis (Figure 8c), for the minimum remained at 1.47 m in both load conditions.

Thus, loading the ankles moved the most stable walk to a lower frequency, with no change in stride length, consistent with previous work showing a decrease in preferred stride frequency. Taken together, those data support the conclusion that an added load shifts the walking attractor along the frequency axis.

**Transition Trials**

Two-way ANOVAs (Load x Transition Direction) were performed on the transition speed, frequency, and stride length.

**Transition Speed**

Averaged over W–R and R–W trials, the mean transition speed decreased slightly from a Froude number of 0.54 (2.19 m/s, $SD = 0.18$ m/s) in the unloaded condition to 0.51
(2.14 m/s, $SD = 0.17$ m/s), that hysteresis effect was not significant, $F(1, 7) = 0.969, ns$; nor was the interaction, $F(1, 7) = 0.31, ns$. To more fully evaluate the hysteresis effect, we investigated individual trends, using a three-way ANOVA (Load x Transition Direction x Participant). That interaction revealed a significant hysteresis effect, as expected, $F(1, 64) = 14.24, p < .001$, but there was also a significant Transition Direction x Participant interaction, $F(7, 64) = 15.88, p < .001$, indicating that there were individual differences in the amount and direction of hysteresis. Analyses of simple effects demonstrated that Participants 2 and 3 showed significant hysteresis ($p < .05$), Participant 8 showed a hysteresis trend (ns), Participants 1 and 5 showed reverse hysteresis trends (ns), Participant 4 showed significant reverse hysteresis ($p < .05$), and Participants 6 and 7 did not have discernible trends (ns). In sum, there were clear individual differences in the direction and amount of hysteresis. We examined this issue again in Experiment 2.

The mean transition speed in each load condition is represented by vertical lines on the fluctuation functions in Figure 8a. Note that the data points rise before the transition, consistent with a weakening of the attractor as speed is scaled away from preferred values and into the usual transition region, thus confirming the results of Diedrich and Warren (1995a) with a finer sampling of speeds.

**Transition Stride Frequency**

Loading the ankles also reduced the mean transition stride frequency, $F(1, 7) = 11.78, p < .05$, which fell from 1.26 Hz ($SD = 0.06$ Hz) in the unloaded condition to 1.23 Hz ($SD = 0.07$ Hz) in the loaded condition. In addition there was a main effect of transition direction, $F(1, 7) = 19.87, p < .01$, such that the W–R frequency (1.19 Hz, $SD = 0.07$ Hz) was lower than the R–W frequency (1.29 Hz, $SD = 0.07$ Hz). That finding is consistent with the jump between attractors illustrated in Figure 3, confirming the results of Diedrich and Warren (1995a). The interaction was not significant, $F(1, 7) = 1.07, ns$.

**Transition Stride Length**

On the other hand, adding a load did not significantly influence the mean transition stride length, $F(1, 7) = 1.11, ns$. Averaged over transition direction, the mean stride length rose only slightly, from 1.74 m ($SD = 0.16$ m) to 1.75 m ($SD = 0.18$ m). There was a main effect of transition direction, however, $F(1, 7) = 19.97, p < .01$, such that the transition stride length was higher in the W–R direction (1.84 m, $SD = 0.17$ m) than in the R–W direction (1.66 m, $SD = 0.19$ m), consistent with the jump between attractors shown in Figure 3. The interaction was not significant, $F(1, 7) = 2.89, ns$.

That pattern of results is summarized in Figure 6. The ankle loads acted to shift the walking attractor along the frequency axis, but not the stride length axis, yielding a corresponding shift in the transition stride frequency without a change in stride length. Thus, as predicted by the theory,
movement of the attractors was accompanied by a parallel change in the walk–run transition.

EXPERIMENT 2
The Effects of Grade

In the second experiment, we attempted to alter the locomotor dynamics by manipulating the grade of the treadmill. As mentioned earlier, existing data indicate that when an individual travels uphill his or her energetically optimal speed decreases, usually accompanied by a reduction in stride length. That finding suggests that the walking attractor moves to a lower speed, largely as a result of a shift along the stride length axis (Figure 3). We tested two grade conditions (0% and 10%), using both plateau trials and transition trials. Our hypothesis was that the most stable walk would shift to a lower speed during uphill locomotion because of a shift along the stride length axis and possibly the frequency axis, inducing a parallel shift in the transition.

Method

Participants
Six people (2 men, 4 women) were paid to participate in the experiment. Individual participants’ characteristics are presented in Table 1.

Apparatus
The apparatus was identical to that employed in Experiment 1. Because ankle weights were not worn, a marker was placed directly over the lateral malleolus (ankle), instead of higher on the shank. Joint angles were measured in the same manner as in the first experiment.

Procedure
The procedure was the same as in Experiment 1, with the following exceptions. First, there were two grade conditions, such that the treadmill was either level (0% grade) or inclined 5.7° from the horizontal (+10% grade). Second, to better assess hysteresis, we tested 15 transition trials in each direction (W–R, R–W) for each condition, rather than 10 trials. That required a third session. In the first session, 30 plateau trials were collected in each grade condition, as before. In the second session, the last 10 plateau trials in each condition were completed, and then the transition trials for one grade condition were performed; the transition trials for the other condition were finished in the third session. The order of the conditions was again counterbalanced across participants.

Data analysis proceeded as in Experiment 1. Data from 14 of the 840 trials collected were missing because of equipment failure or marker occlusion.

Results and Discussion

Plateau Trials
Preferred Stride Length and Frequency Combinations
The preferred combination of stride length and stride frequency at each specified walking speed appears in Figure 9. Two-way ANOVAs (Grade × Speed) demonstrated that both mean stride frequency, $F(9, 45) = 1025.83, p < .001$, and mean stride length, $F(9, 45) = 780.05, p < .001$, increased with speed. There was no effect of grade on stride frequency, $F(1, 5) = 1.02, ns$, or stride length, $F(1, 5) = 0.53, ns$, but the interaction was significant for both frequency, $F(9, 45) = 10.98, p < .001$, and stride length, $F(9, 45) = 10.44, p < .001$. Specifically, at high speeds the mean stride frequency was greater in the uphill condition than in the level condition, and that trend was reversed at low speeds. In contrast, at high speeds the mean stride length dropped in the uphill condition, compared with the level condition, and that trend was reversed at low speeds. Those results are consistent with earlier findings suggesting that grade influences the preferred combination of stride length and frequency, but the nature of the effect depends on speed. The data support the idea that a
FIGURE 9. Preferred stride length and stride frequency combinations in the 0% grade (squares) and 10% grade (triangles) conditions. Large symbols represent the most stable points and mean transition points, with arrows illustrating the direction of the shifts. Dotted lines are iso-speed contours.

FIGURE 10. Mean relative phase of the ankle and hip as a function of speed (Froude number) in the 0% and 10% grade conditions, showing that hip extension led ankle extension. Error bars indicate standard error.
change in grade influences the walking attractor, although because of forced walking they do not indicate the precise direction of the shift.

Mean Relative Phase

Once again, peak extension at the hip led peak plantarflexion at the ankle by about 50° (Figure 10). Watson-Williams F tests for circular variables indicated that there were no overall differences in mean phase as a result of either speed, F(9, 50) = 1.99, ns, or grade, F(1, 10) = 0.64, ns, but the interaction of the two was significant, F(19, 100) = 3.12, p < .01, such that a difference in relative phase between the grade conditions emerged only at lower speeds. Those results indicate that the preferred phase at the walking attractor was altered by the grade manipulation.

Fluctuations of Relative Phase

The standard deviation of relative phase between the ankle and the hip is plotted in Figure 11a, again revealing a U-shaped fluctuation function. An ANOVA (Grade × Speed) confirmed a main effect of speed, F(9, 45) = 14.41, p < .001, with a significant quadratic component, F(1, 5) = 15.84, p < .05. There was no overall effect of grade, F(1, 5) = 3.67, ns, nor was there an interaction, F(9, 45) = 1.20, ns.

Once again, we used Equation 3 to fit the fluctuation curve as a function of speed, stride length, and frequency. Regression constants, \( r^2 \) values, and minima appear in Table 3. The minimum of the fluctuation curve for speed decreased from a Froude number of 0.31 (1.63 m/s) in the level condition to 0.27 (1.53 m/s) in the uphill condition (Figure 11a), a shift of 7.1% of the total speed range in plateau trials. That shift was accomplished primarily via changes in stride length, although stride frequency changed slightly as well. The minimum of the fluctuation curve for stride length dropped from 1.44 m to 1.36 m when participants traveled uphill, a large, 8-cm shift equal to 9.9% of the total observed range (Figure 11c). At the same time, the most stable stride frequency decreased slightly from 1.04 to 1.02 Hz, corresponding to 2.5% of the observed range (Figure 11b).

In sum, these data indicated movement of the walking attractor to a lower speed when traveling uphill, resulting from drops along both control parameter axes, although especially along the stride length axis.

Transition Trials

Two-way Grade × Transition Direction ANOVAs were performed on the transition speed, frequency, and stride length.

Transition Speed

As predicted, the increase in grade reduced the transition speed from a mean Froude number of 0.55 (2.19 m/s, SD = 0.12 m/s) to 0.46 (2.00 m/s, SD = 0.14 m/s), F(1, 5) = 59.88, p < .01, consistent with the shift in the walking attractor. Once again, there was a nonsignificant tendency for the transition speed to be higher in the W–R (2.12 m/s, SD = 0.17 m/s) than in the R–W (2.07 m/s, SD = 0.09 m/s) direction, F(1, 5) = 1.15, ns, and no Grade × Transition Direction interaction, F(1, 5) = 1.57, ns. We used a three-way ANOVA (Grade × Transition Direction × Participant) to more fully investigate the presence of hysteresis. Overall, as predicted,
there was a significant hysteresis effect, F(1, 83) = 18.04, p < .001, but there was also a significant Transition Direction x Participant interaction, F(5, 83) = 15.88, p < .001, indicating that there were individual differences in the amount and direction of hysteresis. Analyses of simple effects indicated that Participants 10, 2, and 3 showed significant hysteresis (ps < .05); Participant 4 showed a nonsignificant trend in the opposite direction, Participant 9 showed a reverse hysteresis effect (p < .05), and Participant 11 failed to show a discernible trend (ns). In sum, by collecting more trials to obtain a better estimate of the W–R and R–W transition speeds, we discovered additional evidence for individual differences in the direction and amount of hysteresis. We return to this issue in the General Discussion.

The mean transition speed in each grade condition, again plotted on the fluctuation function (Figure 11a), demonstrated that fluctuations began to increase before the transition. Similar to the load experiment, those data were consistent with a weakening of the attractor as speed was scaled away from the preferred value and into the typical transition region.

**Transition Stride Frequency**

The change in transition speed was accomplished via changes in both stride frequency and stride length. The increase in grade reduced the mean transition stride frequency slightly, from 1.27 Hz (SD = 0.07 Hz) to 1.24 Hz (SD = 0.06 Hz), F(1, 5) = 11.99, p < .05. The direction of the transition also influenced the mean transition frequency, for the W–R (1.19 Hz, SD = 0.09 Hz) frequency was lower than the R–W (1.32 Hz, SD = 0.05 Hz) frequency, F(1, 5) = 23.70, p < .01, consistent with a jump between attractors in Figure 3. The interaction was not significant, F(1, 5) = 0.78, ns.

**Transition Stride Length**

Grade had a larger effect on the stride length at the transition; the mean transition stride length was 1.73 m (SD = 0.07 m) in the level condition and 1.62 m (SD = 0.06 m) in the inclined condition, F(1, 5) = 29.90, p < .01. The transition direction also had an effect, for mean transition stride length was higher in the W–R direction (1.78 m, SD = 0.05 m) than in the R–W direction (1.57 m, SD = 0.08 m), F(1, 5) = 131.87, p < .001, a finding that is consistent with a jump between the attractors. The interaction was not significant, F(1, 5) = 4.45, ns.

In sum, these data indicate that increasing the grade of the treadmill acted to reduce the transition speed, primarily via a drop in stride length, with a smaller reduction in frequency. As predicted, those changes closely paralleled the shift of the walking attractor (Figure 9).

**GENERAL DISCUSSION**

In the present experiments, further evidence was provided for a dynamic account of gait transitions. In previous research, using transition trials in which participants were allowed to switch gaits (transitory mode), we observed hysteresis, a qualitative change in relative phase, and a sudden jump in relative phase at the transition. In addition, using plateau trials in which participants were held at particular speeds in particular gait (steady-state mode), we observed enhanced fluctuations of phase as the control parameters were scaled from preferred values into the typical transition region. Given the sudden and qualitative change in relative phase at the transition and the weakening of the attractors that were observed as speed was scaled away from preferred values, we took the data as evidence that the transition behaves as a bifurcation between two attractors. In addition, directional shifts in stride length and frequency at the transition were in accordance with the attractor layout of Figure 3 suggested by energetic data. In general, in the present research those observations were replicated. The primary contribution of the observations, however, is that they confirm the prediction that changing the attractor layout by

---

**TABLE 3**

<table>
<thead>
<tr>
<th>Variable</th>
<th>Condition (% grade)</th>
<th>Constant</th>
<th>r²</th>
<th>Minimum</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>A</td>
<td>B</td>
<td></td>
</tr>
<tr>
<td>Froude no.</td>
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<td>0.138</td>
<td>0.925</td>
<td>1.814</td>
</tr>
<tr>
<td>Froude no.</td>
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<td>0.116</td>
<td>0.823</td>
<td>1.575</td>
</tr>
<tr>
<td>Stride freq.</td>
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<td>8.598</td>
<td>3.131</td>
<td>-15.713</td>
</tr>
<tr>
<td>Stride freq.</td>
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<td>6.990</td>
<td>3.056</td>
<td>-12.957</td>
</tr>
<tr>
<td>Stride length</td>
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<td>4.329</td>
<td>-20.425</td>
</tr>
<tr>
<td>Stride length</td>
<td>10</td>
<td>16.601</td>
<td>4.077</td>
<td>-24.910</td>
</tr>
</tbody>
</table>
varying load or grade should induce a corresponding shift in
the transition. Let us discuss this result in more detail.

The Load Manipulation

Loading the ankles was found to affect the walking attrac
tor in Experiment 1: The preferred ankle–hip phase
relationship changed, and the most stable stride frequency
decreased (Figures 7 and 8). That finding indicated a shift
in the attractor along the frequency axis by \(0.05\) Hz, but not
along the stride length axis. The shift was accompanied by
a similar shift in the transition along the frequency axis by
\(0.03\) Hz, with no significant change along the stride length
axis. Thus, as predicted by the dynamic theory, movement
of the attractors to different locations in control parameter
space induced a shift in the walk–run transition in precisely
the same direction (Figure 6). We may represent this change
in the attractor layout resulting from loading as a translation
of the entire layout along the frequency axis (Figure 12a).

In addition to describing the abstract dynamics, it is
important to understand why the load manipulation affect­
ed stride frequency at a biomechanical level. The hybrid
mass-spring/pendulum model (Kugler & Turvey, 1987; Tur­
vey et al., 1988), which has been used to account for the
preferred frequencies of a variety of rhythmic movements,
can also be used to predict the shift in the transition fre­
quency caused by loading. Assuming that each gait pos­
sesses a natural frequency determined by the moment of
inertia of the leg and the (variable) stiffness of muscles and
tendons, and assuming that both stride length and frequen­
cy in the two gaits are approximately equal at the transition,
Diedrich and Warren (1995a) derived an equation for the
transition frequency that minimizes driving force:

\[
\omega_t = \sqrt{\frac{\omega_{t,\text{walk}}^2 + \omega_{t,\text{run}}^2}{2}}.
\]  

For the present data, we calculated the natural frequency in
each gait (\(\omega_{t,\text{walk}}, \omega_{t,\text{run}}\)), following Holt, Hamill, and Andres
(1990), and used Equation 4 to predict the transition fre­
quency. The simple pendulum equivalents of the legs were
computed from each participant’s measurements (Table 1)
and Dempster’s (1955) equations for centers of mass. Ankle
weights were assumed to be point masses located at the
ankles. Muscle stiffness was based on empirical estimates
of the lumped stiffness for each gait (Holt, Hamill, &

In general, the observed shifts in stride frequency were in
the directions predicted by the hybrid model, although the
magnitudes of the actual shifts were not as large as predict­
ed and their absolute values were slightly higher than pre­
dicted. For the attractors, the observed drop in the most sta­
ble walking frequency from 1.03 Hz to 0.98 Hz with a load
was similar to the drop in the predicted natural frequency
from 0.95 Hz \((SD = 0.03)\) to 0.88 Hz \((SD = 0.02)\); the pre­
dicted natural frequency for running fell from 1.43 Hz
\((SD = 0.04)\) to 1.32 Hz \((SD = 0.03)\). For the transition, the
actual drop was 1.26 Hz \((SD = 0.06)\) to 1.23 Hz \((SD = 0.07)\),
whereas the predicted frequency fell from 1.22 Hz \((SD = 0.03)\) to 1.12 Hz \((SD = 0.03)\). Thus, the hybrid model
accounts for the direction of the shifts caused by loading,
consistent with the notion that changes in the transition
result from changes in the natural frequencies of walking
and running. Despite the conceptual insight the model
brings, it does not accurately predict the magnitude of the
shifts and has other limitations because of its linearity (see

The Grade Manipulation

Increasing the grade of the treadmill was also found to
affect the walking attractor in Experiment 2: The preferred

![Figure 12](image-url)
ankle–hip phase relationship changed and the most stable speed decreased, primarily because of an 8-cm drop in the most stable stride length and a slight decrease in the most stable frequency (Figures 10 and 11). Those findings indicated a shift of the attractor along both control parameter axes, and particularly along the stride length axis. As expected, that shift was closely paralleled by a shift in the transition along the stride length axis by 11 cm, and a small shift along the frequency axis (Figure 9). In that case, we can represent the change in the attractor layout because of increasing grade as a rotation of the layout in stride length–frequency space (Figure 12b).

At present, the influence of grade on stride length is not well understood at a biomechanical level. During level walking, the body’s center of mass rises and falls like an inverted pendulum, yielding a conservative exchange of potential and kinetic energy that reduces metabolic cost (Margaria, 1976; McMahon, 1984). When walking uphill, the center of mass rises more and falls less (Minetti et al., 1993, 1994b). It is possible that stride length shortens to preserve a falling phase and maintain the potential–kinetic cycle. For running, Iverson and McMahon (1992; see also McMahon & Cheng, 1990) have recently proposed a mass-spring model in which the muscles and tendons in the leg are represented by a spring that has a different stiffness during shortening than during lengthening. The model assumes that running uphill yields an increase in the shortening spring’s stiffness and a decrease in the lengthening spring’s stiffness, which could influence both stride length and frequency. Although those ideas are suggestive, we do not at present have a biomechanical account of how grade affects the locomotor dynamics.

Stability and Minimum Energy

As noted in the introduction, we believe that the energetics of gait are a manifestation of the underlying dynamics, because they reflect the costs of driving the system away from its attractor states. Although initial evidence appeared to support this hypothesis (Diedrich & Warren, 1995a), in the present experiments we were able to evaluate more fully the relationship between stability and energetics. Unexpectedly, the dense sampling of walking speeds used in this study revealed that the most stable speed (1.66 m/s in Experiment 1, no load; 1.63 m/s in Experiment 2, level treadmill) did not precisely match the energetically optimal speed (~ 1.3 m/s; Margaria, 1938, 1976), although direct comparisons cannot be made across these different studies. In addition, the optimal energy transition speed was slightly different from the actual transition speed when on a treadmill (Hreljac, 1993). In short, energetic measures seem to give slightly different results from the stability measures and the observed behavior. On the other hand, they do correspond more globally in ways that are unlikely to be merely coincidental. For instance, during overground locomotion, there is a jump in speed at the transition that does, in fact, act to minimize energy expenditure (Minetti et al., 1994a). In addition, changes in overall metabolic cost per unit distance closely parallel the U-shaped stability function observed for walking (Figures 3, 8, and 11; Diedrich & Warren, 1995a) and the declining fluctuation function observed for running (Diedrich & Warren, 1995a). Metabolic cost also corresponds to the directional shifts in stability that are seen when grade is manipulated (Experiment 2). In sum, the energy and stability measures closely, but not exactly, correspond to each other. In future research, by collecting both energy and stability measures on the same participants across the range of stride-frequency/length combinations shown in Figure 3 (see Holt et al., 1995, for data of this type), investigators can explore that issue more fully. One possible reason for slight differences between energy and stability measures is that total metabolic energy expenditure is a global measure that includes costs other than simply those associated with driving gait. It is also important to note that there are only small increases in energy expenditure and small reductions in stability when operating slightly away from the optimal values.

Hysteresis

In several previous experiments, significant overall hysteresis has been reported for the human walk–run transition (Beuter & Lalonde, 1989; Diedrich & Warren, 1995a; Hreljac, 1993; Thorstensson & Robertsson, 1987). Diedrich and Warren (1995a) found individual differences in reliable hysteresis, however, because a few participants even exhibited reverse hysteresis, and the present results confirmed that pattern. Two issues need to be addressed with respect to those findings: Individual differences and differences in the study methods used. First, individual differences in the direction and amount of hysteresis such as those reported here are not without precedent. Tuller et al. (1994) also reported individual differences in hysteresis in their investigation of categorical speech perception. They interpreted those findings as further evidence for a dynamical explanation, for they can be accounted for if the system is not strictly deterministic in the transition region (as a result of the presence of multiple attractors and noise). If that is true, then one may observe enhanced contrast (reverse hysteresis), identical transition speeds (critical boundary), and hysteresis, for all of those relationships can occur in a bistable region because of the action of noise-related perturbations. The fact that we find all of those patterns in human gait is also indicative of bistability, such that two gaits are possible in the transition region. The specific reason why one participant will show an overall hysteresis effect whereas another will not remains a question for future investigation.

Second, a possible reason for differences between studies in reported hysteresis is that there were small differences in the measurements of the transition speed. For instance, although Hreljac (1993) found evidence for significant hysteresis, he varied speed stepwise in a sequence of plateaus and defined the walk–run transition as the first running plateau and the run–walk transition as the first walking
plateau (e.g., Hreljac, 1993). That measure of transition speed would generate a hysteresis effect even if the gait transitions occurred between plateaus at identical speeds. Thus, it is likely that Hreljac’s results were biased toward finding hysteresis. In addition, Thorstensson and Robertsson (1987) found that the degree of hysteresis may depend on the rate of acceleration of the treadmill, which leads to differences in the time scale of change in the control parameter (\(t_{control}\)). We are currently investigating such differences in methods by running participants in several types of transition trials (Diedrich & Warren, 1995b, in press). Surprisingly, our preliminary results indicate that significant hysteresis is actually more likely when the control parameter is scaled continuously, and less likely when the approach to the transition is step-like, consistent with differences in control parameter scaling and transition definitions underlying discrepancies in reported hysteresis results. Once again, such a relationship can be expected from a dynamic perspective because when the approach to the transition is step-like, the long plateaus in the transition region may allow for the observation of spontaneous shifts between gaits that result from a decrease in the system’s equilibration time. Those spontaneous shifts would eliminate any strong hysteresis effects.

Steady-State and Transition Methods

In both the present study and our earlier work (Diedrich & Warren, 1995a), we used a combination of steady-state and transition methods. The strengths of those two types of methods and related modeling approaches are currently a topic of debate in the literature (Fuchs & Kelso, 1994; Schmidt & Turvey, 1995). Our claims regarding the dynamical nature of the transition between walking and running rest on both types of data. We argue that the combination of a sudden and qualitative change in relative phase observed in the transitory state as well as a weakening of the attractor as the control parameters are scaled during steady states offers support for the theory that the gait transition behaves as a bifurcation between two relative phase attractors. Regardless of these concerns, however, our plateau (steady-state) trials were run in ascending and descending steps, with each plateau lasting 40 s at a particular speed. Because those procedures adhered to the delay convention, it is likely that our measures of enhanced fluctuations in plateau trials would be similar to measures of critical fluctuations measured in transition trials. It is possible, however, that the instruction to remain in a specified gait altered the potential landscape. We plan to investigate such intentional effects in future research by measuring critical fluctuations in transition trials.

Conclusions

The present results lend support to the dynamic theory that gait transitions behave as bifurcations between attractors. We tested the hypothesis that changes in the attractor layout yield corresponding changes in the gait transition. Increasing load or grade shifted the walking attractor to different stride length and stride frequency combinations, which induced precisely parallel shifts in the walk–run transition, consistent with the theory.

An important implication of that finding is that the attractor layout is determined by the dynamics of the task. Those task dynamics incorporate both the dynamics of the action system, including the moment of inertia of the limbs, muscle stiffness, and perceptual information about forcing or cost, as well as the dynamics of the environment, including support surface characteristics, external loads, and related perceptual information. The evidence indicates that attractor states and critical states emerge from the interactions of the total system. Thus, any analysis of preferred gaits and transitions should consider the task dynamics, rather than simply the dynamics of the locomotor system in isolation from the environment.

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NOTES

1. Technically, Diedrich and Warren (1995a) did not measure critical fluctuations, but instead measured fluctuations of phase in steady states (see General Discussion).

2. Those values come from our regression fits of Margaria’s (1938, 1976) energy data, in which we used equations of the form of Equation 3, taken from Inman, Ralston, and Todd (1981; for similar results see Bobbert, 1960).

3. See Note 2.


5. Note that although we did not measure critical fluctuations per se, those procedures adhered to the delay convention. It is therefore likely that our measures of enhanced fluctuations in plateau trials would be similar to formal measures of critical fluctuations (see General Discussion).

6. A continuous measure of relative phase was not employed because the joint motions were not strictly sinusoidal. The use of continuous relative phase on these data would therefore have yielded uninterpretable measures.

7. In the rare event of the mechanical failure of the foot switches, we calculated the transition variables in accordance with the method employed for Group 1 in Diedrich and Warren (1995a).

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